The Impact of Surface and Geometry on Friction Coefficient Behavior of Artificial Hip Joints

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Abstract:

Friction coefficient tests were conducted on 28-mm and 36-mm-diameter (Ø) hip joint prostheses of the four different material combinations, with or without the presence of Ultra High Weight Molecules Polyethylene (UHWMPE) particles using a novel pendulum hip simulator. The effects of three micro dimpled arrays on prosthesis head against a polyethylene cup and a metallic cup were investigated. Clearance played a vital role in friction coefficient of various geometrical and materials oriented artificial hip joints. Micro dimpled metallic heads yielded higher friction coefficient against polyethylene cup, however, these dimpled arrays were significantly effective in friction coefficient reducing to metal on metal prosthesis. The *in situ* images disclose the evidence of enhanced film formation due to the dimple, a main mechanism that contributed to reduced friction.

Keywords: Hip joint prosthesis, geometry of prosthesis, micro dimple, friction coefficient, lubrication film thickness

1. Introduction:

The number of hip replacement procedures is more than 2 million per annum worldwide, which will increase twofold by 2020 because of increasing aging population [1,2]. According to the 2014 Canadian Joint Replacement Registry, the numbers of hip and knee replacements have increased by 16.5% and 21.5%, respectively, over the last five years[3]. The demographics include an increasing number of younger patients (45-64 years) and thus, prostheses are now required to last over 30 years[3-5].

The orthopedics device manufacturing companies, equipped with advanced manufacturing facilities, are able to produce prostheses with high precision and very smooth surface (10-20 nm). Although ISO standard is followed during *in vitro* testing, revision rates of prostheses in clinical application are as high as 8-10%, within an average 15-year lifespan per device, according to the UK and Canadian national survey reports[3,4]. Surgical techniques and precision, as well as patient-related physiological factors and activity levels are important determining factors for the survival rates of implanted hip or knee joints[6]. In addition to these factors, excessive wear rate and its associated debris are the major drivers of revision surgery[7,8]. Wear is a complex mechanism, involving lubrication and coefficient of friction (REF- Puccio-biotribology). Scholes et al, showed that lower friction factors are associated with thicker lubrication film formation. It is noted that articular cartilage along with synovial fluid, play the key role in protecting the joint interface from mechanical wear and facilitating a smooth motion (friction coefficients can reach as low as 0.001) [30, 31]. Thus Friction Coefficient (FC), wear rate and their associated debris are the main measurable outcomes to evaluate the performance of newly developed hip prostheses.

One of the widely-used methods of determining FC is based on strain gauge, principally suitable to pin-on-disk tribo-tester (non-conformal contact)[9]. A hip joint is a ball-and-socket joint with conformal contact. Hence, it is challenging to measure FC and real-time wear rate, using a strain gauge. Thus, the measurable outcomes for determining FC, using hip simulators, are generally focused on the evaluation of wear rate only[10,11]. The pendulum hip simulator is a novel device[10] that can be used to obtain real-time velocity profile and estimated frictional coefficient data of hip prosthesis, while accurately maintaining prosthesis geometrical configurations, such clearance and sizes.

Prosthesis size and clearance are important selection criteria for hip replacement[12,13]. A 28 mm-diameter (Ø) hip prosthesis is widely used in many countries[14]. Despite of previous studies[13,15] have shown that hip joint prostheses with a larger diameter have lower wear rate, their fundamental tribological behavior and lubrication viscous effect are not yet clearly understood. Similarly, no single study has investigated the effect of size and clearance on FC with all of the major material pairs, such as metal-on-metal (MoM), ceramic-on-ceramic (CoC), metal-on-polyethylene (MoP) and ceramic-on-polyethylene (CoP) prostheses. Similarly, in the current literature, Only a few articles mentioned the effect of wear debris in third body abrasive wear mechanism and a possible reduction by surface modification[16,17]. Micro-dimpling is a method of surface modification used in many engineering applications, such as in engine cylinders[18,19], ball and rolling bearings[20,21] and has potential in orthopedics[22,23]. Micro-dimpling can increase hydrodynamic pressure by enhancing film formation; it acts a reservoir to, simultaneously, increase lubrication and entrap debris.

However, a synovial-lubricated micro *dimpled-conformal contact* (i.e. the bearing in hip and knee joints with defined size and clearance) is not yet fully understood. Eleven articles have been published investigating the effect of surface texturing on orthopedic implants; six considered 'MoP' [16,17,23-26], one 'CoC'[22], and the remaining four, 'MoM' implants [27-30]; Four patents have been granted based on these endeavors[31-34]. The dimple parameters utilized were diverse, but most of them demonstrated that a micro-dimpled surface reduced friction and wear. Among the MoM prosthesis, two are theoretical and other two are based honed surfaces. Therefore, a micro dimple surface technique is yet to be investigated in metal-on-metal hip joints.

In this study, four types of FC experiments were conducted, including a) a 28-mm Ø and a 36-mm Ø prosthesis with four material combinations (MoM, MoP, CoC and CoP); b) the impact of polyethylene particles in the MoP hip joints interface; and c) the effect of three micro dimple arrays on MoP and MoM prostheses.

2. Material and Methods

2.1. Materials: Hip joint prostheses with diameters of 28- mm and 36- mm Ø and in four material combinations, including MoM, MoP, CoC and CoP, as detailed in Table 1, were obtained from University Hospital Olomouc, Czech Republic. The 36- mm Ø MoM had the largest clearance (609 µm) and the 28 mm Ø MoM has lowest clearance (274 µm). Clearance and contact surfaces were measured, using a 3D optical scanner active fringe projection (GOM ATOS Triple Scan, GOM mbH, Germany). The accuracy of the measuring system is justified according to the acceptance test (VDI/VDE 2634 - Part 3) whereas the probing error size is ±0.005 mm. The Post-processing calculate on the diameter of the spherical surface of prosthesis was performed, using the ATOS professional software. The exact diameter of contact surface of artificial glass cups were 28.08 and 38.08 mm. Artificial femoral heads (details in the table 1) were delivered in the original package from the manufacturer and the exact diameter after scanning was 27.988 and 35.299 mm.

UHMWPE powder ET306010 (Goodfellow Cambridge Ltd, Huntingdon, UK) was used to test the influence of debris on wear rate. To identify a reliable circumference value, we measured 5 random particles, 3 times each. The particles were first sputtered on the microscope slide and placed into the profile-meter. Three images were captured from different slide orientations and analyzed, using a 3D optical microscopy (Contour GT-I, Brucker, Italy). The edges in all the images were identified and broken grains were discarded. The particles supplied had an average equivalent circumference of 150 μm, but the deviation was large, 100 μm.

Three types of micro dimples arrays (square, triangular and circular) were fabricated on the 28 Ø Aesculap, Isodur (metal) prosthesis heads. Indentation techniques[21] were used to produce the square and triangle arrays, and micro drilling was performed to create circular dimple arrays. A 3-axis rotation axis device was used to produce dimples on the prostheses because of their spherical shape. The indentation has one axis of rotation (Z) and two axis planner movement facilities, which allowed the fabrication of the square and triangular micro dimple arrays on the top 6x6 mm^2 area of the prosthesis. The outer ring dimples were unavoidably slightly tilted from the perpendicular position. The tip of the indenter was tilted at an angle of 15^o from the vertical axis. The CNC micro drilling machine, with three rotational axes, enabled fabrication of dimples perpendicularly to the prosthesis head. The tungsten carbide WC drill bit of 200μm Ø was used to produce the circular dimple arrays. The dimple parameters are presented in Table 2, and the 3D images are shown in Fig. 1. The images and their roughness profile were captured and evaluated, using a 3D optical microscopy (Contour GT-I, Brucker, Italy). The images showed presence of protuberance formations around the dimples, which were removed by a further surface finishing, using 0.5 micron diamond paste. The dimple roughness profiles before and after the polishing are shown in Fig. 2.

2.1 Friction test

Tribology tests were performed, using a novel pendulum hip joint simulator (Fig. 3), described by Choudhury et al[35,36] . The experimental parameters are shown in Table 3.

The prosthesis head is loaded by two handed weight bar which can be loaded and unloaded manually. The pendulum is pulled up to 16º, and then released to swing freely. The frequency and the initial peak amplitude were 0.5 Hz and 0.9 radians, respectively. The amplitude reduced to zero over time due to the FC. The measured angular velocity profile was transferred to the computer and FC was calculated from the recorded values of angular amplitudes. The applied load was 2 KN, simulating three times body weight during a standard walking gait for a MoC hip joint[37,38]. Bovine serum (25%) was used as lubricant, and temperature was maintained at 37° C to simulate body temperature.

2.3. Computational simulation:

A numerical analysis (ANSYS R.15) was carried out to investigate the deformation of softer UHWMPE against dimpled metallic head. The numerical model consisted of two planar bodies. The dimpled surface was set on the bottom with fixed mode and non-dimpled cup was loaded with contact pressures of 10 MPa and 24 MPa. The dimple was 250 µm in diameter and 40 µm in depth, which corresponded to the dimple geometry of the circular array.

3. Result and Analysis

3.1 Effect of Prosthesis Size on Friction Coefficient:

The effect of prosthesis size and clearance on FC of MoP and CoP prostheses are shown Fig. 4. — FC of the 36-mm-Ø MoP prosthesis started from 0.12, increased gradually, and finally settled to 0.18 after ten test repetitions in steps of 3 minutes. FC of the 28-mm-Ø MoP prosthesis started from 0.14 and gradually increased to 0.19 after ten repetitions. Therefore, the FC trends of the MoP hip prostheses of different sizes were very similar. MoP prosthesis size did not have noticeable influence on FC. FC of the 36-Ø-CoP prosthesis started from 0.095 for the first experiment and extended to 0.11 at its 10^{th} repetition, which was lower than that of 28-mm-Ø CoP (13% lower at the first experiment and 22% after the 10^{th} repetition). Overall, CoP FCs are much lower compared to MoP FCs, regardless of their sizes.

MoM and MoP prostheses had similar FC trends. FC was not influenced by the prostheses of different sizes (28-mm and 36-mm Ø). However, the maximum FC of MoM (0.16) was lower than that of MoP (0.19). On the other hand, CoC exhibited a different FC profile to that of MoM. FCs for the 28-mm and 36-mm-Ø CoC started from 0.1 and 0.09, respectively, and extended to 0.14 and 0.12, respectively, at the tenth test repetition. The final FC of the 36-mm-Ø CoC prosthesis was 16% lower than that of the 36-mm-Ø one. The average FCs of MoM and CoC prosthesis are shown in the Fig. 5 and 6, respectively.

3.2 Effect of Particle Size on Friction Coefficient

Fig. 7 illustrates the impact of artificially induced UHWMPE debris on MoP hip joint prosthesis. At the very beginning of the experiment, the added particle helped reduce FC; FC was slightly less (3%) than that of the prosthesis interface ‘without debris’. The difference reduced to zero at 5th experiment. However, FC profiles of prostheses with added particles started to exceed those of the prostheses with no-added particle from the 6th experiment, reaching a difference of 6.5% at the 10th replication. The post-experiment image of the prosthesis cup (Fig. 3) revealed a layer of deposited PE particles, which confirmed the presence of particles at the interface during the experiment. A large number of particles were also found to attach to the prosthesis head.

3.3 Effect of Dimples on Metal on Polyethylene Prosthesis Friction Coefficient

Prosthesis heads with square, circular and triangular dimple arrays were tested against polyethylene acetabulum to determine the resulting FCs (Fig. 8). FCs of all the dimpled prostheses were higher than those of the non-dimpled pairs. For example, the FCs of square and circular arrays were 0.13 at the beginning of the experiment, 13% higher than that of the non-dimple prosthesis. Moreover, the difference increased to 17% at the end of 10th experiment. On the other hand, prostheses with triangular array started with similar FC values as those of non-dimple prostheses, but their FCs gradually increased to 20% at the end of the 10th test, compared to the non-dimpled MoM. Ito et al[25] reported improved frictional coefficient by using similar pattern of dimple. However, our study revealed that large and deep micro dimples, regardless of their arrangements, did not decrease FC, but instead, increase FCs compared to non-dimpled MoP prostheses. From the computation simulation, it has been revealed that an elastic deformation occurred to UHWME, as a consequence, the edge of the deformed UHWME penetrated into the dimple. The depth of the penetration were 8.2 μm and 3.2 μm under contact pressure 24 MPa and 10 MPa, respectively (Fig. 9). Off Couse, the deformation happened due to the lower modulus of elasticity of the UHWME compare to that of CrCoMo. The penetrated edge of UHWME are responsible for the increased FC.

3.4 Effect of Dimples on Metal on Metal Prosthesis Friction Coefficient

Fig. 10 presents FC trends of dimpled MoM prostheses. FC of MoM prosthesis with square arrays started from a much higher magnitude of 0.17 for the first test, compared to that of non-dimpled prosthesis (0.13), but gradually decreased with increasing number of repeated tests. However, the FC profile of MoM prostheses with non-dimple arrays increased slightly, with an increasing number of repeated tests. The FC of MoM prosthesis with square array reduced by 22% compared with non-dimpled Isodur-Aesculap, and by 54%, compared with non-dimpled Metasul-Zimmer MoM hip prosthesis.

The frictional profile for the MoM prosthesis with triangular dimple array was very similar to the one for the MoM prosthesis with squared dimpled array, although its magnitude was slightly higher. While FCs for MoM implants with square and triangular arrays improved with increasing number of tests, the FC for MoM implant with circular array increased substantially, reaching almost 40% higher than the non-dimpled sodur-Aesculap and Metasul-Zimmer hip prostheses after the 5th test repetition. It is to be noted that the dimples for the circular array had larger depth and diameter than those for the square and triangular arrays (Table2) because of the manufacturing difficulties. The velocity profiles also confirmed that the squared dimple array had a viscous damping effect, an indication of better lubrication formation over the contact zone[35]. Therefore, we performed a preliminary in-situ observation of square array in order to observe the lubrication film behavior. To do this, the metal cup was replaced with a glass-made cup. The applied load was reduced in magnitude (500 N), but other experimental conditions remained unchanged, including temperature, lubricant and frequency. From the in-situ observation, a clear enhanced film formation was identified, which was believed to be the main reason for lowered friction coefficients of dimpled prosthesis. The enhanced film formation (70nm) is illustrated in Fig. 11.

4. Discussion

Friction coefficient, wear rate and film thickness are the three main fundamental aspects of Biotribology[31,39]. Interface properties and prosthesis geometry are the key areas that can influence the lubrication mechanism of artificial hip joints and, consequently, fluctuation in friction coefficient and wear rate[9,32]. In this study, an investigation was conducted on the frictional behavior of different implant material combinations, such as MoM, MoP, CoC and CoP hip joints prostheses, which are manufactured by two well-renowned medical device-manufacturing companies─Zimmer and B. Braun. Hence, the clearance and surface roughness were well defined and consistent. Other experimental conditions, such as load (2.5 KN), temperature (37^{o}C), swing angle 32^{0} degree and 25% bovine serum, were accurately replicated to the clinical conditions of implanted hip joints[13,31,40].

FC of the 28-mm-Ø CoP prosthesis was significantly higher than that of the 36-mm-Ø for every test repetition. FC for both CoP head sizes increased with increasing number of test repetitions. The FC trend was similar to those of the MoP prostheses. Interestingly, the FC of CoC for the similar geometrical combinations (28-mm-Ø & 36-mm-Ø) are similar trend as per as CoP, however, the scenario changed —there is no significance FC difference between 28-mm-Ø & 36-mm-Ø combinations. Clearances of the 36-mm Ø and 28-mm Ø CoC prostheses were 562 µm and 363 µm respectively (difference: 199 µm); those of the 36-mm Ø and 28-mm Ø MoM prostheses were 609 µm and 274 µm (difference 335 µm), respectively. Therefore, the difference of Clearances of the prostheses is the main factors that influence their frictional behaviors. Using a simplified film thickness equation, Dowson[13] showed that, when diameter is larger and clearance are minimum the maximum film thickness yield. According to their derived theory, the elastohydrodynamic film thickness (h_{min}) between perfectly smooth surfaces is directly proportional to (diameter, mm)$^{2.19}$ and inversely proportional to (clearance, µm)$^{0.77}$. Using this equation, the calculated ratio (diameter$^{2.19}$/clearance$^{0.77}$) were 18.37 for the 36-mm-Ø MoM hip prosthesis, 33.98 for the 28-mm-Ø MoM, 19.54 for the 36-mm-Ø CoC and 15.78 for the 28-mm-Ø CoC. These values show that, despite its larger diameter, the 36-mm-Ø MoM cannot improve the prosthesis frictional behaviour unless they have a justified clearance. Similar mechanisms happen to the present experiment, the CoP hip prosthesis as well and larger diameter CoP had lowered FC values. Larger-diameter hip prostheses are reported to have lower risks of dislocation[12-14]. This study reveals that a larger-diameter hip prosthesis, with a minimum diametrical clearance, can improve frictional behaviour.

It is reported that third-body abrasive wear[16,17], which usually involves fine debris generated from the rubbing area, have lower influence in friction and wear; third body abrasive wear rate is ten times slower than second body abrasive wear[9,32]. However, the slow progression of third body abrasive can decrease the surface fatigue strength which could lead to implant failure over time[32]. In this study, we did not find any significant impact of polyethylene debris on frictional behaviour of MoP prosthesis. Although there was clear evidence of debris at the interface, these could have deformed elastically or plastically due to the high contact pressure, thereby reducing their impact upon wear. At the 6th repeated experiment, there were small differences between the FC of hip prostheses with ‘no particle’ and those with ‘added particles’ and the differences gradually increased with repeated number of tests. This could be because of an increased number of particles produced at the interface with time.

Fabricated micro dimples are one of the advanced surface modifications for enhancing lubrication, controlling contact pressure and wear debris[17,33]. In this study, micro dimples were found to adversely affect the resultant frictional behaviour of MoP prosthesis, which contradicts findings in the current literature[17,23]. Ito et al[25] stated that a concave pattern dimples with 0.5 mm diameter, 1.2 mm pitch, and 0.1 mm depth fabricated on the Co-Cr alloy head could significantly reduce friction and wear rate when rubbing against UHMWPE. Ito and co-investigators' findings[25] indicate that concave dimpled MoP could be a good solution for preventing excessive UHMWPE wear rate, however, they also suggested that further testing on the dimpled profile should be conducted , using a standard hip simulator. Sawano et al[16] suggested not to use micro dimples deeper than 1 μm since these were found to result in higher friction and wear, however, they used pin-on-disc specimens, which was not a true replication of artificial hip joints in the terms of geometry and clearance. Although our experiments lasted for only 30 minutes, our results showed evidence that the edge of the deformed UHWME penetrate onto the micro dimple of CrCo head were mainly responsible for the increased FC. The numerical analysis (Fig. 9) revealed that height of the UHWMPE deformed edge could be around '4 μm at the 10 MPa contact pressure' and '8 μm at the 24 MPa contact pressure' into the dimple profile. These contact edges could act as an interlock and increase the FC.

The most important finding of this study was the effect of micro dimples on MoM prostheses. A significantly lower and steady FC profiles were obtained for micro dimpled MoM prostheses, compared to those of non-dimpled MoM prostheses. A clear evidence of enhanced film formation was found, however, we did not measure the quantitative values of film thickness at this stage. The evaluation of film thickness during various sliding and rolling conditions are currently being investigated. The stable and reduced FC of the square and triangular dimple arrays could have a significant impact on the long-lasting stabilization of the hip joint. The circular dimple array, however, resulted in an increased FC. This could have been due to the larger diameter and depth of the circular array, which could have reduced lubrication distribution and film formation. An *in situ* observation would help understand the lubrication distribution and their associated film formation; this is currently under investigation by our investigational team. The square array showed better performance than the triangular array because of the consistent line-up to the direction of rolling of the prosthesis head. It is to be noted that the rolling of the pendulum hip joint is a unidirectional swing, which is slightly different from the physiological movement of the natural hip joint, but very similar to that of the knee joint. Therefore, the simulator needs to be adapted to simulate the multidirectional movement of the hip joint to obtain an optimum dimple arrangement for hip joints in clinical usages.

There is a lack of investigations on conformal contact of hip joint FC. Most of the published studies report FC values, based on pin- or ball-on-disk (non-conformal) with either unidirectional or reciprocating motion[15,41]. Thus there has been a wide variations in reported FC values (0.03-1)[41]. Scholes et al.[15] investigated on the frictional behaviour of commercially-available total hip joint prosthesis, however, they provided frictional factor values, which was similar in magnitude to FC values, but varied with the pressure distribution over the head. Moreover, they concentrated only on 28-mm diameter prosthesis, without evaluating the clearance. Dowson et al.[13] confirmed that prosthesis clearance and size had huge influence on wear mechanism, however, they did not provide their friction coefficient outcomes. This study has explored FC of the two most clinically-utilized sizes and material pairs of hip prostheses, along with their respective defined clearance[14,39]. Furthermore, this study has included advanced and well-defined micro dimpled surfaces that resulted in substantial reduced and stable FC. Results of this study could therefore be beneficial to clinicians and prosthesis manufacturing technologists.

5. Conclusions

This study consisted of three major parts: a) investigating frictional behavior of two sizes and four material combinations of commercially available hip joint prostheses, b) the influences of artificially added UHWMPE particles at the MoP interface and c) the fabrication of three dimple arrays on spherical hip prosthesis heads and their tribological impact. The conclusion of the student can be pointed out as follows:

- Size and clearance influence the tribological behavior of hip joint prosthesis. A 36-mm-Ø CoP, with a clearance of 453 μm, yielded a significant reduction in FC of 22%, compared to a 28-mm-Ø CoP. However, the 36-mm-Ø MoP and 28-mm-Ø MoP hip joint prostheses had similar FC values because of their respective clearance values. FCs of the MoM prostheses had similar trends, but were 12% lower for the 36Ø CoC (clearance 562 μm), compared to the 28-mm-Ø CoC (clearance 363 μm). The predicted film thickness and experimental viscous signs supported the friction coefficient results.
- The artificially added UHMPE particles had a minimal effect on the frictional behavior of the 28-mm-Ø Metal on Polyethylene prostheses. The softer UHMWPE particles were found to be deformed under the contact pressure. Post-experiment images revealed the presence of UHMWPE particles at the cup and head interface.

- The 28-mm-Ø MoP prostheses with three micro dimple arrays yielded a significant increase in FCs, compared to that of the non-dimpled MoP. The edges of ‘elastic-deformed UHMWPE cup into rigid micro dimpled prosthesis head’ were thought to be the cause of the increased FC. Dimples with smaller diameter and depth are believed to produce lower FCs.
- The square dimple arrayed MoM prosthesis (28mm Ø) yielded a significant friction reduction of 24% and 35%, compared to that of non-dimpled Brown and Zimmer MoM prostheses, respectively. These corresponding percent reductions for the 28-mm-Ø MoM prosthesis with triangular dimple arrays were 19% and 30%, respectively.

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Fig1. Fabricated dimple arrays before polishing: a) square b) triangular c) circular; and after polishing: d) square e) triangular f) circular

Fig 2. Surface roughness profile of dimple before and after the polishing

Fig 3. Pendulum hip simulator

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